AC/DC Millivoltage Sensor by means of ITO-coated Optical Fibers: Towards Monitoring of Biosignals

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***Abstract*—This contribution shows the monitoring of AC and DC millivoltage signals by means of lossy mode resonances generated by Indium Tin Oxide (ITO) on optical fibers. Sensors were obtained by sputtering ITO thin-films onto 25 mm-length segments of 200 μm bare optical fibers. Depositing a 1 μm thin-film of ITO leads to obtain reduced thin-film resistances of near 340 ohms. This allows the detection of voltage signals by monitoring the wavelength shift of the resonances. Sensitivities up to 40 nm/V can be achieved when tracking sinusoidal signals of a few cents of mV peak-to-peak. This opens the path for further research pursuing the detection of biomedical signals.**

Keywords—fiber-optic sensors; lossy mode resonances; indium-tin-oxide (ITO); thin-film; AC/DC voltage.

I. INTRODUCTION

Monitoring biosignals is becoming crucial nowadays. A continuous improvement of the state of the art is carried out to register, for example, proper ElectroCardioGram (ECG) or ElectroEncephaloGram (EEG) signals, to adequately diagnose human diseases. In this sense, the current technology capable of detecting the variation of these signals is based on specific electrodes which are distributed throughout certain locations on the body, thus obtaining the desired electrical signals coming from inside the organism.

On the other hand, today it is common to find sensing applications where optical fibers are used. Among them, temperature [1], strain [2] or bending [3] are good examples of physical parameters that can be detected with this platform. Moreover, it is also possible to detect chemical and biochemical reactions, just by depositing the proper nanomaterials onto these waveguides. In this sense, optical fibers can be used as biosensing platforms [4], but they are quite common in medicine already, due to their use in endoscopy or in varicose veins and prostate cancer surgery [5], among others.

These capabilities can be achieved since optical fiber presents some benefits that make it be suitable for being used in medicine. Some of them are biocompatibility, reduced size, flexibility, weight and cost, capability to work within complex matrices media, such as the blood [6], capability of allowing multiparameter sensing [7] and diversity of light detection configurations.

Given these interesting characteristics, the main purpose of this contribution is to use optical fiber as an optical electrode and then show a first approach to the detection of biosignals. However, a major drawback is found when trying to perform such experiments: the optical fiber itself presents electromagnetic interference (EMI) immunity [8]. The reason is that the frequency variations of the electromagnetic fields surrounding the fiber are several orders of magnitude below the light frequencies propagated inside of it. Therefore, before proceeding to measure, it is necessary to prepare the fibers in order to detect electrical signals.

Up to date, there are several ways to make optical fibers be sensitive to electrical variations. Among them, the use of electro-optic tunable Bragg gratings [9] or interferometers [10] is quite common. The strategy consists of depositing nanostructured materials in, on or around the fiber, which can interact at the same time with the outside DC voltage variations and with the light propagating inside the fiber [11].

However, when dealing with biosignals, some aspects should be considered. Among them, the low voltage values obtained (a few mV peak-to-peak), varying with a certain frequency and with measurements carried out in water-based media, like the sticky gels used in electrodes to obtain EEG/ ECG signals. As commented, electrodes have practically solved this issue nowadays, since they are an effective way to discern the actual electric activity from electrochemical processes inside the organism. Therefore, the challenge is to design an optical technology that may enhance or, in any case, complement, the results provided by current electrode-based technologies.

The way to address this issue in this contribution follows the strategy of a recent work by *Smietana et al.* [12]. The idea is that by depositing thin films of ITO around an optical fiber it is possible to generate Lossy Mode Resonances (LMRs) capable of detecting electrochemical reactions as if the fiber was an optical electrode. The interest of that contribution is the good correlation between the wavelength shift of the obtained LMRs and the electrochemical phenomena taking place in the working solutions. From a different point of view, the fact of being capable to translate DC voltage signals into LMR wavelength shifts using saline solutions is the perfect excuse to try varying the wavelength shift of such LMR as a function of either DC and even AC voltage signals as if they were produced inside the

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organism. This contribution will show an idea of how it could be done.

II. Materials and Methodology

A. Sensors manufacture

The first step was to generate the sensing phenomena, this means, the LMRs. To this purpose, several optical fiber segments from a 200/250 µm (core/cladding diameter) cladding removable multimode optical fiber (CRMMF) from Thorlabs Inc. (FT200EMT) were obtained. They were stripped, uncladded and then immersed in detergent, deionized water (18.1 MΩ/cm), piranha solution (30% H_2O_2 + 70% H_2SO_4) and again deionized water during 10 minute-time periods. The aim of these steps was to remove the covers, clean the fibers surface and prepare them for the ITO deposition.

A sputtering deposition using an ITO target from ZhongNou Advanced Material Technology Co. was then performed over the fibers. For this purpose, a K675XD sputtering equipment from Quorum Technologies, Ltd. was used with an Argon partial pressure of 5.5×10^{-2} mbar and a current intensity of 140 mA. In order to make a study on the spectrum evolution as a function of the deposition process as well as on the coating thickness, sputtering times of 1.5, 4, 6 and 13.5 minutes were programmed.

B. Experimental set-up

Once the sensors were fabricated, they were connected to an optical set-up similar to that depicted in Fig. 1. Basically, 25 mm-length segments of the ITO-coated fibers were cleaved at 90º and then spliced to two CRMMF pigtails. A broadband light source (Alphabright Inc.), covering a wavelength range from 400 to 1800 nm, launched light into one end of the new created waveguide. The other end was connected to a spectrometer (XR1-FLG OceanOptics Inc.), which permitted to monitor a wavelength range from 200 to 1000 nm.

The DC/AC voltage introduction was performed by either a power supply or a waveform generator (Agilent 33250A). Generators were connected to the fibers through a couple metal bars fixed to them using a conductive glue as shown in Fig. 1. Then, the sensors were immersed in low molarity saline solutions composed of sodium, potassium and calcium chlorides dissolved in deionized water. This clearly increases the conductivity of the set-up and, at the same time, tries to imitate the internal conditions of neurons, as indicated in [13].

Regarding the optical data processing, the obtained transmission spectra were captured during the experiments and referenced to the spectrum obtained when the sensor was immersed for the first time in the saline solution. Additionally, in order to track the shifts in wavelength experienced by the LMRs during the whole process, a MATLAB® algorithm based on least squares was programmed.

III. Results and Discussion

A. Characterization of the ITO thin-films

Once the fibers were deposited with the sputtered ITO, they were sent to an SEM microscope to characterize their thickness. Also, a polymeter was used to check their resistivity in air before proceeding to the detection in the neuron-mimic solution. As it can be observed in Fig. 2, the first LMR showed up after 1.5-minute deposition time, what involved an average thickness of 155 nm and a resistance of 1 MΩ. The second LMR showed up after depositing a thickness of near 505 nm. The registered coating resistance was 291 kΩ. The values for the third LMR were a thickness of 886.5 nm and a resistance of 37.6 kΩ. Finally, obtaining the 4th, 5th and 6th LMRs after 13.5-minute deposition involved a 926.5 nm-thick coating and a resistance of just 5 kΩ. It is simple to deduct that as the thickness of the material deposited onto the fiber increases, the more effective area it has to conduct electricity. According to the well-known expression of a conductor resistance (R = ρl/A), the fact of increasing its transversal area is consistent with a reduction of its resistance value.

Additionally, by comparing the resistance values in air and in the saline solution (see Table 1), it can be observed that the fact of immersing the sensor in the solution clearly decreases the resistance of the ITO coating. In these conditions, there are additional free ions that can transport the charges. Consequently, if a voltage signal is introduced, it is easier to move these charges along the ITO-coated fiber and thus, track the wavelength shift of the generated LMRs, as it will be shown later.

B. Refractometric and voltage measurements

In view of the very low resistance presented by the 1 µm-thickness ITO thin-film deposited onto the optical fibers (4th LMR), these samples were used to make the rest of the measurements. First, some refractive index measurements were taken by immersing the sensors in glycerine solutions of 0, 15%, 30%, 50% and 70% in water. Maximum sensitivities of 500, 304 and 187 nm/RIU were achieved for LMRs 4 to 6. Since the 4th LMR was located in NIR wavelengths (see Fig. 3) its sensitivity was the highest among the 3 LMRs showing up in the spectrum. This is something typical in LMRs, as indicated in [14].

TABLE I. ITO coating resistances for LMRs 2 to 6

Surrounding medium	Coating resistance (kΩ)		
	2nd LMR	*3rd LMR*	*4th, 5th & 6th LMRs*
Air	291	37.6	5
Saline solution	7.5	3.1	0.342

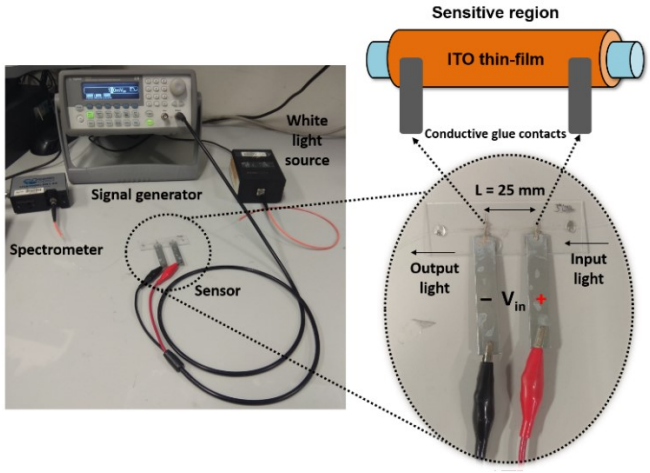

Fig. 1. Experimental set-up to carry out the experiments.

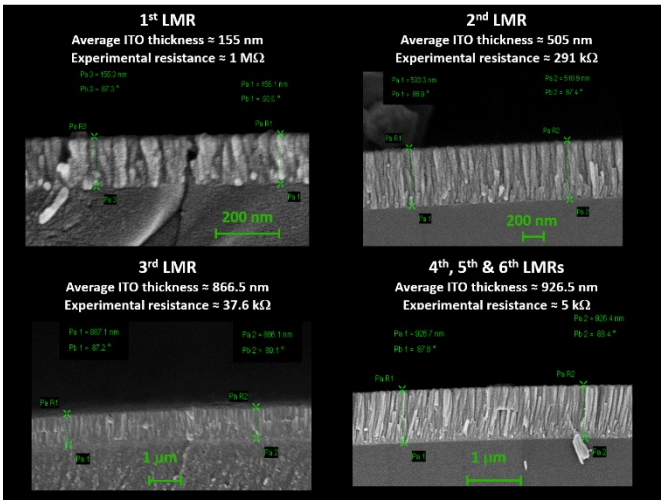

Fig. 2. SEM images to characterize the ITO nanocoating thickness as the LMRs show up in the spectrum. The corresponding experimental resistance for each coating thickness is also indicated.

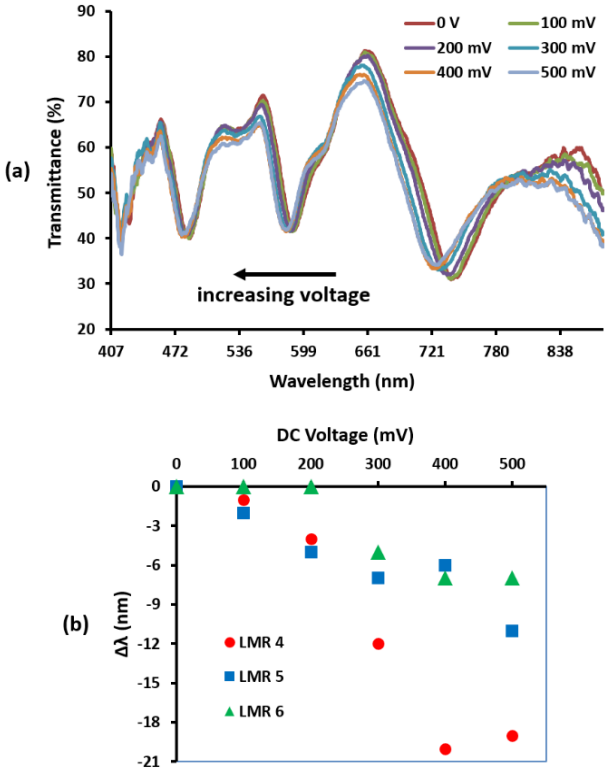

Fig. 3. Spectral behavior of 4th, 5th and 6th LMRs when the sensor is subjected to voltages from 0 to 500 mV. (b) Relative shifts for LMRs 4 to 6 as a function of the increasing DC voltage.

After checking the sensitivity to changes in the surrounding medium of the sensor, the detection of DC and AC millivoltage signals is addressed. In this sense, Fig. 3a and 3b show the spectral behavior of the LMRs as a function of the increasing DC voltage between 0 and 500 mV. LMRs experience a blue-shift to lower wavelengths and the sensitivities to voltage variations are 40, 22 and 14 nm/V respectively for LMRs 4 to 6.

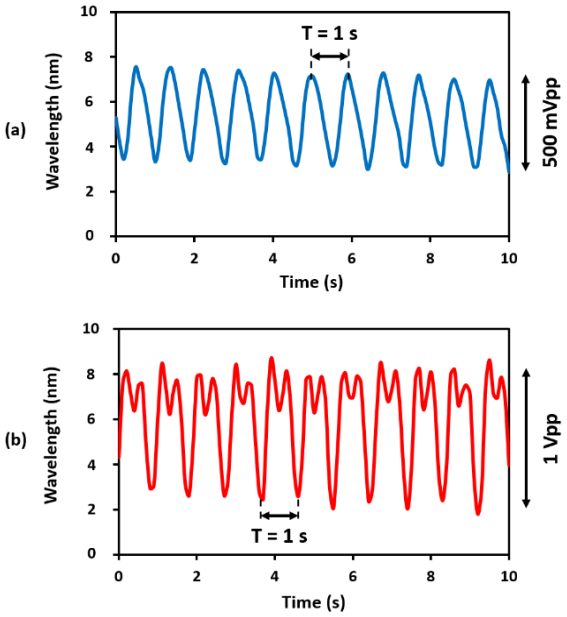

Fig. 4. Monitoring of the stationary regime of sinusoidal signals of 1 Hz frequency and peak to peak amplitudes of 500 mV and 1 V respectively for the 4th LMR. Notice that the higher the amplitude, the more the LMR shifts in wavelength.

In the end, the fact of being located at higher wavelengths makes the fourth LMR be more sensitive to the outside changes, so the higher is the voltage drop, the more it shifts. The same can be observed in Fig. 4 when introducing peak-to-peak sinusoidal signals of 500 mV and 1 V respectively with no offset. This can be explained by a variation in the electroactivity of the ITO surface when subjected to these experimental conditions, as indicated in [12,15]. Particularly, when the voltage increases, the ITO thin-film is reduced, since the electrons are captured by the free cations dissolved. Consequently, the effective refractive index of the modes propagating inside the waveguide is reduced and that is why the LMR blue-shifts. On the opposite, when the voltage decreases, the ITO layer is oxidized, allowing a better conduction and, therefore, red-shifting the LMR.

IV. Conclusions

To sum up, an optical electrode based on Lossy Mode Resonances (LMRs) generated with ITO has been developed and characterized, not only in DC, but also in AC regime. The thicker is the ITO layer, the lower resistance it presents, so it is easier to induce voltage drops that provoke refractive index changes which can be detected. As an electrode, the working principle is based on redox reactions between the ITO layer and the surrounding medium: a solution that mimics the internal conditions of neurons.

This research opens the path to monitor biosignals which characteristics and testing are similar to those studied in this contribution. In this sense, the next steps are: improve the sensitivity of the LMRs to electrical signals, test their response to higher frequencies and, in a future, try to apply these enhancements to the detection of electrical signals as an additional technique to diagnose neural or cardiac malfunction.

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